A FAST ALGORITHM FOR PHASED ARRAY IMAGE RECONSTRUCTION IN MAGNETIC RESONANCE IMAGING

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Abstract: Radiofrequency receiver coils in magnetic resonance imaging systems are used to pick up the signals emitted by the nuclei. Surface coils provide a high signal-to-noise ratio because of their small sensitive region but the usable field of view is also limited to the size of the sensitive region. Using coil array permits to obtain high SNR and a large region of sensitivity: the outputs from the receiver channels are combined in order to construct a single composite image from the data of many coils. For the image construction, usually sum-of-squares (SoS) method is used, which combines data without the knowledge of the coils sensitivity but it is known to provide low contrast images. In this work we investigate and test on MR images a simple method (SUPER algorithm) which uses an estimation of coils field maps to combine the data from the phased array elements to yield an image with higher contrast respect to the usual SoS.

Introduction

Radiofrequency (RF) coils are key components in Magnetic Resonance Imaging (MRI) systems. In order to obtain high quality MRI images, RF coils should be able to generate wide field of view (FOV) with high RF magnetic field homogeneity in transmission and to achieve high signal-to-noise ratio (SNR) in reception.

Many different RF coils have been designed and according to their shapes, they can be categorized into two groups. The first group is called “volume coils” (Helmholtz coils, saddle coils and birdcage coils) and these coils are often used both for transmission and reception. The second group is called “surface coils”, which include loop coils of various shapes. These coils are usually much smaller than the volume coils and, hence, have higher SNR because they receive noises only from nearby regions. However, they have a relatively poor field homogeneity and, thus, are mainly used as receive coils.

The main problem that arises with surface coils is that the usable FOV is limited to the size of the sensitive region, whereas it is desiderable to have a large FOV because the region of interest is often not known prior to the first scan.

A solution is to use a coil array to provide a large region of sensitivity. Each coil is connected to an independent receiver channel and the outputs from these channels are combined in an optimum manner with a phase correction dependent on the point in space from which the signal originated. In this manner, we can obtain the high SNR of surface coils and a large FOV usually associated with large coils. Such a coil array is named “phased array” and its use in MRI was firstly described in 1990 by Roemer [1]. A good summary of this technology is provided in the review paper [2].

Phased array MRI requires an image reconstruction algorithm to combine the individual coil images into a single composite image with full FOV.

The most commonly used method for image recostruction is the so-called “Sum-of-Squares” (SoS) method, in which each pixel value of the reconstructed image is the square root of the sum of the squares of the pixel values corresponding to the individual coils in the array [1]. This method has the advantage that detailed RF field maps of the coil do not have to be known but it is known to introduce signal bias in the estimated image, causing a decrease in image contrast, because the pixels are not weighted by the RF field produced by each coil.

For increasing contrast, image data from arrays should be combined pixel by pixel with each coil's contribution weighted by its local sensitivity. But direct calculation of the individual coil field maps requires precise a-priori knowledge of each coil’s position in relation to the image FOV and this requirement is particularly problematic for flexible phased-array coils.

In this work we tested a method for the estimation of the coil sensitivity directly from each coil image and then we used these RF maps for optimal image reconstruction with SUPER algorithm [3]. We also compared the resulting image with the SoS reconstruction, giving a measure of the benefit of performing SUPER reconstruction in terms of image contrast increase.

Materials and Methods

The image S_i obtained from coil i in an array of N coils, neglecting the noise output from the coil, is given by:

$$S_i = B_i \cdot \rho \qquad (1)$$

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where ρ is the MR signal acquired from an ideal homogeneous coil and B_i is the spatial sensitivity of coil i. Assuming no correlations between the coils, the optimal estimate for ρ is obtained from [3]:

$$P_{opt} = \frac{1}{\sum_i B_i^{\,2}} \sum_i S_i B_i \quad (2)$$

with i=1..N. Since the B_i are not usually known for each pixel, it is possible to use an estimate of the coil sensitivity in the form:

$$B_i^{'} = \frac{S_i}{\sqrt{\sum_k S_k^{\,2}}} \quad (3)$$

where k=1..N. Substituting $B_i^{'}$ from Eq. (3) for B_i in Eq. (2) gives the SoS reconstruction:

$$S_{SoS} = \sqrt{\sum_k S_k^{\,2}} \quad (4)$$

SoS is the state-of-art technique for signal combination from phased array coils. It combines the data without knowledge of the coils fields and at the same time, generally, preserves the high SNR of the coils. As described in Eq. (4), each pixel value of the reconstructed image is the square root of the sum of the squares of the pixel values corresponding to the individual coils in the array.

If all coils in the array have similar noise and each image has a high pixel SNR, SoS method results in a high SNR. In the case of coils with different noise, we initially developed and tested a method for preliminary equalizing the background noise of the separate images.

The background noise equalization was performed using a weight for the pixels of each image: this weight (hereafter referred to as “equalization ratio”) is equal to the ratio between the background noise standard deviation of the image with the lower noise level and the background noise standard deviation of each image.

For example, if a 2-element phased array coil provides two images with σ_1 and σ_2 values for the standard deviations and supposing that $\sigma_1 < \sigma_2$, the equalized SoS combination provides:

$$S_{SoS} = \sqrt{S_1^{\,2} + \left(\frac{\sigma_1}{\sigma_2} \cdot S_2\right)^2} \quad (5)$$

where if the images have equal noises ($\sigma_1 = \sigma_2$), Eq. (5) becomes the classical SoS reconstruction. The extension for a N-element phased array coil provides:

$$S_{SoS} = \sqrt{(S_{\min})^2 + \left(\frac{\sigma_{\min}}{\sigma_1} * S_1\right)^2 + \ldots\ldots\left(\frac{\sigma_{\min}}{\sigma_{N-1}} * S_{N-1}\right)^2} \quad (6)$$

where $S_{\min}$ is the image with lower background noise standard deviation $(\sigma_{\min})$ and S_i are the images with σ_i as value of standard deviation. Fig. 1 shows a phantom obtained using a 2-coil phased array.

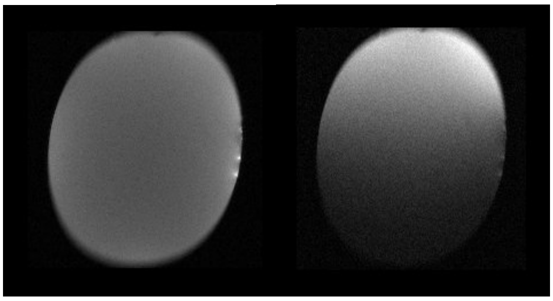

Figure 1: MR images of a phantom acquired with a 2-element array coil

The background noise standard deviation measurement provides, respectively, σ_1=33.45 for the coil 1 and σ_2=59.92 for the coil 2. The equalization ratio is $r = \sigma_1 / \sigma_2 = 0.56$. In Fig. 2 we can note the differences between the SoS reconstruction without equalization (on the left) and the equalized SoS reconstruction (on the right): this last has a lower noise and a higher homogeneity respect to the simple SoS.

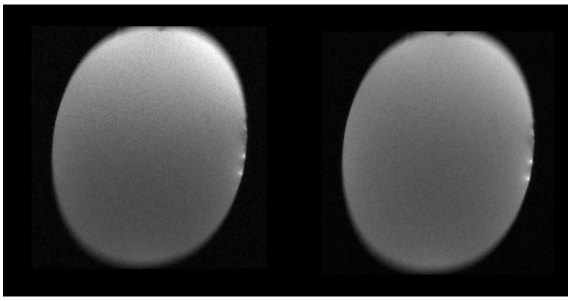

Figure 2: SoS and equalized SoS reconstructions

However, a disadvantage for a SoS image is related to a decrease in image contrast, because the pixels are not weighted by the RF field produced by each coil.

For increasing contrast, image data from arrays should be combined pixel by pixel with each coil's contribution weighted by its local sensitivity. As shown in Eq. (1), the signal from each coil can be expressed as the product of the native MR signal with the coil sensitivity. Because the sensitivity generally vary slowly across the imaging volume, we verified that good estimation of the coils field maps can be obtained with a smoothing procedure applied directly to the image spatial domain, convolving each image with a lowpass filter F_i, as according to the following equation:

$$B_i^{'} = S_i \otimes F_i \tag{7}$$

where $\otimes$ denotes convolution operation and $B_i^{'}$ is the coil sensitivity estimate.

Results

We initially designed and tested a smoothing filter described as:

$$F_i = \begin{cases} \frac{1}{w}\sum_{j=0}^{w-1} S_{i+j-w/2} & \text{i = w/2,...., N - w} \\ S_i & \text{otherwise} \end{cases} \tag{8}$$

where S_i is the MR image matrix, w is the filter width and N is the number of elements of S_i. If the width of the smoothing window is wide enough, the smoothed images are a good estimate of the coils sensitivity maps because, as described previously, $S_i \to B_i$.

For the test of the designed filter and the reconstruction method which will be described later, experimental data were acquired on a vertical B_0 MRI system produced by Esaote Biomedica (E-Scan 0.18T, open MRI dedicated to musculoskeletal limbs studies) using a 2-element array coil for the imaging of the shoulder. Each element is a 8cm radius circular loop and the two loops are placed near perpendicular each other and mutually decoupled.

We used these parameters for the T_1-weighted Spin-Echo imaging sequence: TE=26msec, TR=600msec, slice thickness=4mm, FOV=13x13cm, number of signal averages=1, pixel dimension=0.7 mm.

Fig. 3 shows the two 128x128 images obtained with the MR scanner.

Using a 50x50 smoothing filter, we obtained the estimated coils sensitivity maps with a good agreement with the expected field pattern (Figure 4).

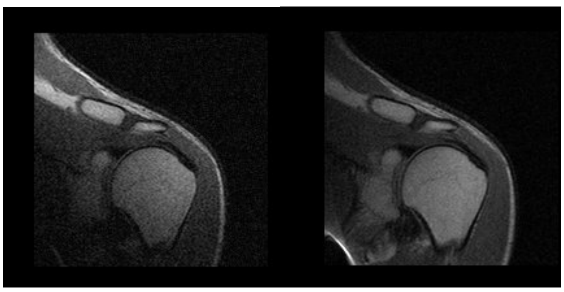

Figure 3: MR images acquired with a 2-coil array

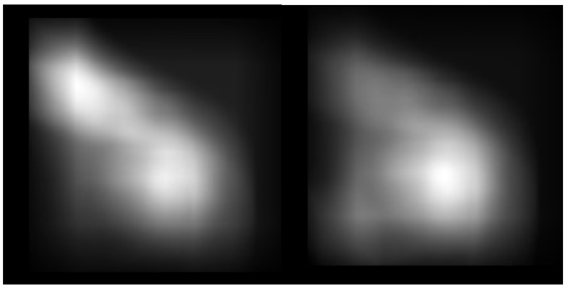

Figure 4: Smoothed MR images

However, in the smoothed images we noted the presence of artifacts localized in the image pixels corresponding to signal fast transition areas.

To minimize these artifacts, we tested a Hanning window for the extraction of the coils sensitivities from the images, defined as:

$$W(k) = 0.5 - 0.5 \cdot \cos(2\pi k / N) \tag{9}$$

where k=0, 1,..N-1 and N is the window width.

In Figure 5 are showed the images obtained from the convolution of a 60x60 Hanning window with the MR images. Again, the filtered images are a good estimate of the RF coils maps but the artifacts have become negligible.

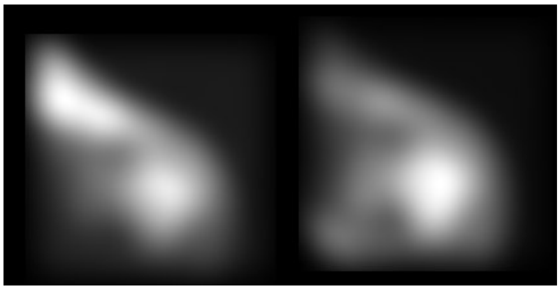

Figure 5: Hanning filtered MR images

After applying the filter to uncombined images and estimating the coils sensitivities, we can perform the SUPER reconstruction [3]:

$$S_{SUPER} = \sqrt{\sum_i B_i^{'2}} \left| P_{opt} \right| \tag{10}$$

where

$$P_{opt} = \frac{1}{\sum_i B_i^{'2}} \sum_i S_i B_i^{'} \tag{11}$$

Substituting Eq. (11) in Eq. (10), the optimal reconstruction for a 2-element phased array coil is given by:

$$S_{SUPER} = \frac{S_1 B_1^{'} + S_2 B_2^{'}}{\sqrt{B_1^{'2} + B_2^{'2}}} \qquad (12)$$

where S_1 and S_2 are the images from coil 1 and coil 2, and $B_1^{'}$ and $B_2^{'}$ are the estimated sensitivities of coil 1 and 2. Obviously, before applying SUPER algorithm we equalized the background noise of the S_i images.

Fig. 6 shows a conventional SoS combination (on the left) and the SUPER reconstruction (on the right) of the two separate images of Fig. 3.

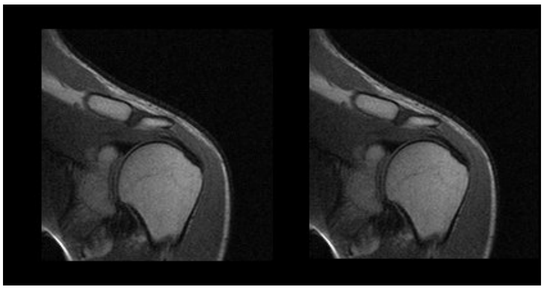

Figure 6: SoS and SUPER reconstruction

Comparing SoS and SUPER reconstruction we noted that this latter has much improved contrast, with a darker background. To give a measure of the benefit of performing this SUPER combination respect to a conventional SoS, we measured the difference between the means of the pixel intensities calculated in a region-of-interest outside the object being imaged, providing both an estimate of the background noise decrease and image contrast increase. The result is a good 5% gain in using this SUPER instead SoS reconstruction.

Discussion

Roemer [1] showed that for optimal SNR, image data from phased arrays should be combined pixel by pixel with each coil's contribution weighted by its sensitivity at that location. However, the knowledge of coil sensitivity is not generally available for each pixel and so SoS combination has been adopted as the state-of-art technique for signal combination from phased array coils. This method is known to introduce signal bias in the estimated image, causing a decrease in image contrast, because the pixels are not weighted by the RF field produced by each coil.

SUPER algorithm permits to combine the image data from arrays with each coil's contribution weighted by its local sensitivity. We demonstrated that coils sensitivities can be estimated directly from the spatial domain of the image using a smoothing procedure, because the sensitivity generally vary slowly across the imaging volume. This operation was performed convolving each image with a lowpass filter.

We verified that, respect to the usual SoS reconstruction, the application of SUPER algorithm with the coils sensitivities estimated using a Hanning window directly from the images of each coil permits to increase contrast between high and low signal regions, which improves the object definition.

In order to quantify the degree of the contrast improvement of the SUPER reconstruction in comparison with the conventional SoS, we may calculate the difference $\delta = S_{SoS} - S_{SUPER}$, providing [3]:

$$\delta = \frac{N-1}{SNR^2} \qquad (13)$$

where it is evidence that the benefit of performing SUPER instead of SoS combination is inversely related to the local SNR and increases with the number of the coils. This becomes increasingly important for phased-array with a large number of elements.

Conclusions

An efficient algorithm for the combination of image data from array coils, where each signal is weighted by the coils sensitivities, was presented. Using a Hanning window applied to the images of each coil, we estimated the RF coils maps with good accuracy before applying SUPER reconstruction algorithm which allows contrast image increase respect to the application of conventional SoS.

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